EXPERIMENTS IN MOVING TOWARD A FEASIBLE STRIP-PET SCANNER

An Undergraduate Research Scholars Thesis

by

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Submitted to the Undergraduate Research Scholars program
Texas A&M University
in partial fulfillment of the requirements for the designation as an

UNDERGRADUATE RESEARCH SCHOLAR

Approved by
Research Advisor: Dr. Mahapatra

May 2016

Major: Physics
Mathematics

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ABSTRACT

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For many years, PET scanning devices have been effective in many areas of medicine, particularly in functional imaging and detection of cancer. However, the most common designs are costly, and could be made more efficient. Recently, a new design has been proposed, involving longer detectors and uncommonly used scintillating materials. We are building and testing small examples of this type of setup, in the hopes of determining accuracy, viable methods of operation, and overall feasibility.

ACKNOWLEDGEMENTS

I would like to thank my research advisor Dr. Mahapatra for making this thesis possible. I would also like to thank all the members of the research team for the countless hours dedicated: Jaime Cardona, Josh Flores, Reece Goldsberry, Justin O'Connor, and Robin Snellings. Finally, I thank William Baker for helping give us direction throughout the project.

CHAPTER I

INTRODUCTION

One of the most visible advances in medicine in recent times has been medical imaging. The ability to produce images of the inside of a patient has been a great boon to discovering and diagnosing diseases inside the body which would have otherwise required exploratory surgery or may have never been found. Among the first of these technologies to be developed was that of x-ray imaging, starting with the discovery of x-rays by Röntgen in 1895. As technology, and computers specifically, became more advanced, many new and safer ways of imaging became available. Methods such as Computed Tomography (CT), Ultrasound, and Magnetic Resonance (MRI) are used widely today to help identify and understand a variety of diseases and conditions.

Among the methods developed in the twentieth century was the field of Nuclear Medicine. Nuclear Medicine is based on principles involving the usage of relatively weak sources of radiation which are introduced into the body of the patient.[1] This is commonly done through the use of radiopharmaceuticals, which are organic substances (e.g. sugars) that have been made radioactive. When these radioactive substances decay, they release ionizing radiation, which can be picked up by detectors outside the body and reconstructed into an image.

Positron emission tomography

One of these methods of imaging is called a Positron Emission Tomography scan, or PET scan. For a PET scan, a radioactive tracer consisting of a positron (β^+) emitting radiopharmaceutical is

introduced into the body, usually via injection. When a positron is emitted, it collides with an electron within the body and annihilates it, converting entirely into energy. This produces two collinear (back-to-back) gamma rays, which are then detected by the PET scanner. In the most common scanner design, the most significant data that any detection is able to give is the line that the gamma rays traveled on, sometimes called the line-of-flight (or LOF), which is given by the position of the detectors which they triggered. The quantity of annihilation events detected by a given pair of detectors gives the average density of the radioactive substance along that line, and with this information given on every measured LOF, a density can be assigned to every point in the scanner. This is the local density of the radiopharmaceutical inside the patient, and can be interpreted as a three dimensional image.[2]

This method has some very practical advantages. Importantly, PET scans are a commonly used method of functional imaging. Because sugars are a primary source of energy in the body, they tend to collect more heavily in parts of the body which are active. Because the radiopharmaceutical is itself a sugar, this means that the density of radiation detected with a PET scan is directly related to how active a certain area of the body is. For this reason, PET scans are commonly used in the study of the function of the brain, and how it responds to certain stimuli. They are also used in the study and diagnosis of various degenerative brain conditions, such as dementia and Alzheimer's disease. The same property also makes PET quite effective in detecting certain cancers. Cancer, by its nature of being a mass of uncontrolled cell division, has a very high metabolism. Thus, the tumors show up strongly on a PET scan.

PET limitations

Unfortunately, PET scanners are limited by their cost. For instance, the most commonly used isotope in the creation of the radiopharmaceuticals, ^{18}F, has a half-life of approximately 109 minutes.[3] This is too short to distribute or store the materials like other types of indicators. Instead, the PET detector must be accompanied on-site by a cyclotron, a type of particle accelerator, in order to create the radiopharmaceuticals on demand.[3] The design of PET detectors can also be a cost barrier to many hospitals. One of the most common designs is that of a ring of detectors placed around the patient. This ring is densely covered on the inside with individual detectors which create a flat image of the region inside the ring. The entire ring is then moved along the patient in order to generate the three-dimensional image of the tracer density. Each detector in the ring is usually a *block detector*. Each block detector consists of two parts: the scintillator matrix, and an avalanche photodiode (APD).

A scintillator is a material (usually, a crystal) which, when struck with a form of radiation, absorbs the radiation and releases light. In PETs, they absorb the collinear gamma rays and turn them into visible light. The scintillator matrix is an array of scintillator crystals, or a single crystal with an array of cuts to make each section optically independent. Inorganic scintillators are common, specifically bismuth germanate (BGO). An avalanche photodiode (APD) is a device, usually made from a semiconductor, which absorbs light and turns it into a current. This current would act as a positive detection in a PET scanner. In a block detector, the scintillator matrix is connected to a 2x2 array of APDs. Ideally, the block detector can differentiate between detections in the four different sections of the scintillator matrix, but this is not perfect.[2]

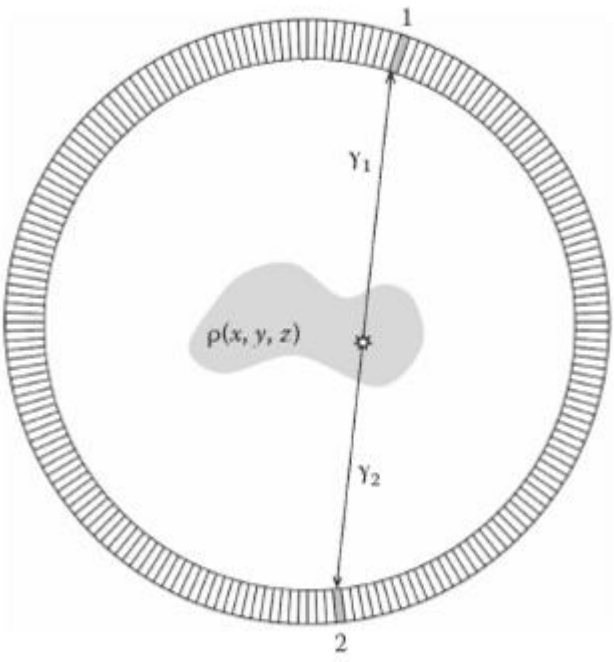

Figure 1: Standard setup for a PET detector ring. The collinear gamma rays, γ_1 and γ_2, are detected by the detectors on the ring, and through many such detections, the density of radiopharmaceutical, ρ, can be determined.[2]

The focus of research

The primary focus of the research behind this paper is a proposed PET scanner redesign, called the strip-PET.[4] In the strip-PET design, a series of long scintillator crystals lie along the length of the patient, instead of facing toward them. Two detectors (either APDs or photomultiplier tubes (PMTs), which accomplish the same goal) are placed at the ends of each of the long scintillating crystals. When a gamma ray is absorbed by the crystal and the crystal scintillates, it sends a pulse down the crystal which can be detected at either end. Data about the two detections, such as the intensity of the light detected or the time delay between detections, can be used to determine where along the crystal the scintillation occurred. Given that two scintillators are struck by the collinear gamma rays, the LOF can be determined, and the tracer density can be calculated in the same ways as the standard PET. If this method can be refined, it could be more efficient than and as accurate as commercial PET scanners in use today. One of the primary advantages of this design is the cost

of the scintillators. Almost all PET scanners in commercial use are made with inorganic scintillators. The strip-PET design, however, can make use of much cheaper organic scintillators instead.[4] The design doesn't use a moving ring in its detection, which can decrease the time necessary to construct an image; it can also detect LOFs which are skewed along the length of the patient that otherwise wouldn't have been in the detecting plane of the detection ring, and missed by the more common design. And we believe that this design could be integrated into other detectors, such as MRI, for more detection options, or simultaneous detection.

In this paper, we focus on experiments conducted with the goal of developing a method for accurately reconstructing the position of a source given the readings of detectors on a long scintillating crystal. We apply standard statistical methods to determine the accuracy of our data samples, and try to determine faithful models for the interactions of the radiation and the light inside the crystal in hopes for the most accurate position reconstruction available to us.

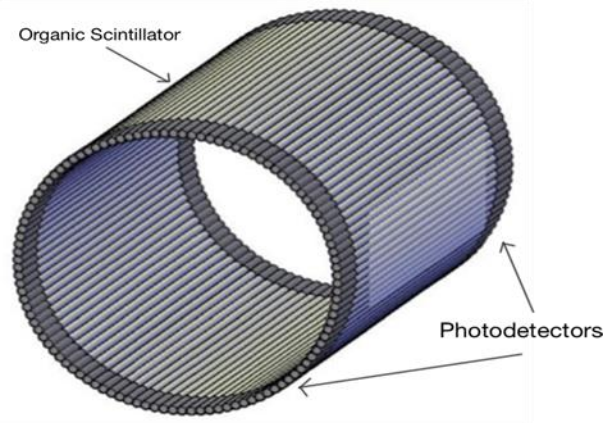

Figure 2: Visualization for the proposed strip-PET detector[4]

CHAPTER II:

METHODOLOGY

The research was conducted as a team, and mostly took place in the summer of 2015. The team included Jaime Cardona, Josh Flores, Reece Goldsberry, Justin O'Connor, Robin Snellings and myself.

For our research, we utilized a simple setup similar to the one described above. Building a full scale experiment is outside the scope of this research, and so a small scale model was used in testing the feasibility of the full scale version. The feasibility testing was primarily concerned with the accuracy of determining the position of a radiation source given the detection data collected over a period of time. For the first experiment, we set up two strip detectors and used a small amount of ^{22}Na as our radiation source. ^{22}Na is a common source material for PET test objects; it has a half-life of 2.6 years and releases annihilation photons with an energy of 511 keV. The ^{22}Na was collimated using two bricks of lead so that the interactions in the scintillators would be happening at specific positions along the crystals.

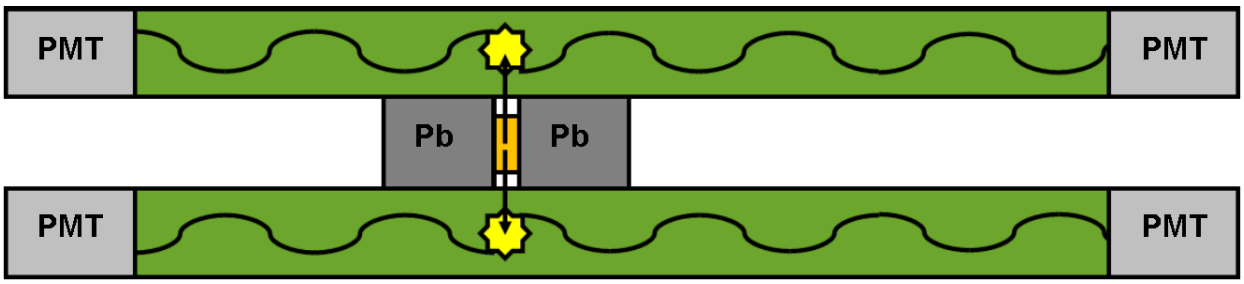

Figure 2: Diagram of our experimental setup. The ^{22}Na source is between the lead bricks

Setup

Scintillators

As stated, the primary property of a scintillating crystal is that it absorbs high energy radiation and converts it into visible light. The time between absorption of the radiation and the emission of the visible light depends on the material the scintillator is made of, but usually happens on the scale of nanoseconds. In general, there are two types of scintillating crystals: organic and inorganic.

For the small scale experiments, we focused on using organic scintillators and their properties. One of the main advantages of a strip-PET design is that it can utilize the cheaper organic crystals instead of the inorganic crystals used in contemporary PET imaging. We decided to use the plastic organic scintillator Bicron BC 408, made by Saint-Gobain Crystals. It was relatively easy to procure in large amounts, and had a decay time of around 2.5 ns. For the experiments, we used two scintillators which measured 18 cm in length. They were cylindrical, with a diameter of approximately 2 inches. The crystals were wrapped end to end with electrical tape so as to cancel the effect ambient light would have on the detection and emission of photons by the scintillator.

PMTs

At the ends of the scintillators, we placed a photomultiplier tube to detect the emitted visible light. A photomultiplier tube is a device which detects very low intensity light by creating an electrical impulse upon detection. When light enters the tube, it strikes a photoelectric material and releases a few electrons. These electrons are accelerated by an electric field, and are used to release more

electrons. At the other end of the tube, enough electrons have been accelerated to create a small current, detectable by an electronic device.

The PMTs were labeled A through D, with A and C being attached to one scintillator, and B and D being attached to the other. The PMTs were attached using electrical tape, wrapped tightly to prevent ambient light from affecting the readings of the detectors. We also placed optical grease between the crystal and the PMT to lessen the effect of the transition between the scintillator and the detector. This helped prevent light from being reflected by the interface.

Oscilloscope

The electronic device we used to detect this current was a Pico 4824 PicoScope Oscilloscope. This device is able to record the voltage created by the current to a time accuracy of milliseconds. This was enough to record the impulse and determine important characteristics like amplitude and FWHM (Full Width at Half of Maximum, a measure of the spread of the impulse). The data from a recorded impulse was saved and imported to a computer, where we processed it using a MATLAB program. With this, we could find the characteristics of a large number of impulses and determine various properties of their distribution, like the average maximum or the standard deviation of the maximums.

Recording data

During a test run, we set the PicoScope to record on the order of 2000 impulse events. The radiation source would be placed at a specific distance along the scintillators, and the PicoScope would be set to only record events where at least two PMTs detected an event simultaneously. This is what

would be expected if the source sent back-to-back radiation into both of the scintillators, and helped us rule out other detection events, like cosmic rays. Once a distribution of the events was made in MATLAB, we moved the source to another position, and created a plot of the average energy of the events versus the position of the source. With this graph, we could then place the source at a position, and determine its location by finding the energy of the recorded events. This would tell us how accurate our position reconstruction could be.

CHAPTER III:

RESULTS

Much of our research was done with the understanding that future students would continue the project with better and more advanced equipment. Thus, the focus was on finding possible issues with these types of setups, and to determine the best methods and programs to move forward with. We focused on a parameter which we called the *discriminatory value* or the *rho-value*, which is a normalized difference in the energies detected by the PMTs on either end of a scintillator. The rho-value is defined as

$$\rho := \frac{E_1 - E_2}{E_1 + E_2}$$

With E_1 and E_2 being the energies recorded by the PMTs on a single scintillating crystal, i.e. A/C or B/D. We measured this value at 15 points along the scintillators, at 1 cm intervals from 3 to 18 cm along their length. A graph of the rho-value versus source position, as well as a linear regression, is shown in Figure 3. Our results show that the rho-value correlates very strongly with the position of the source along the scintillator.

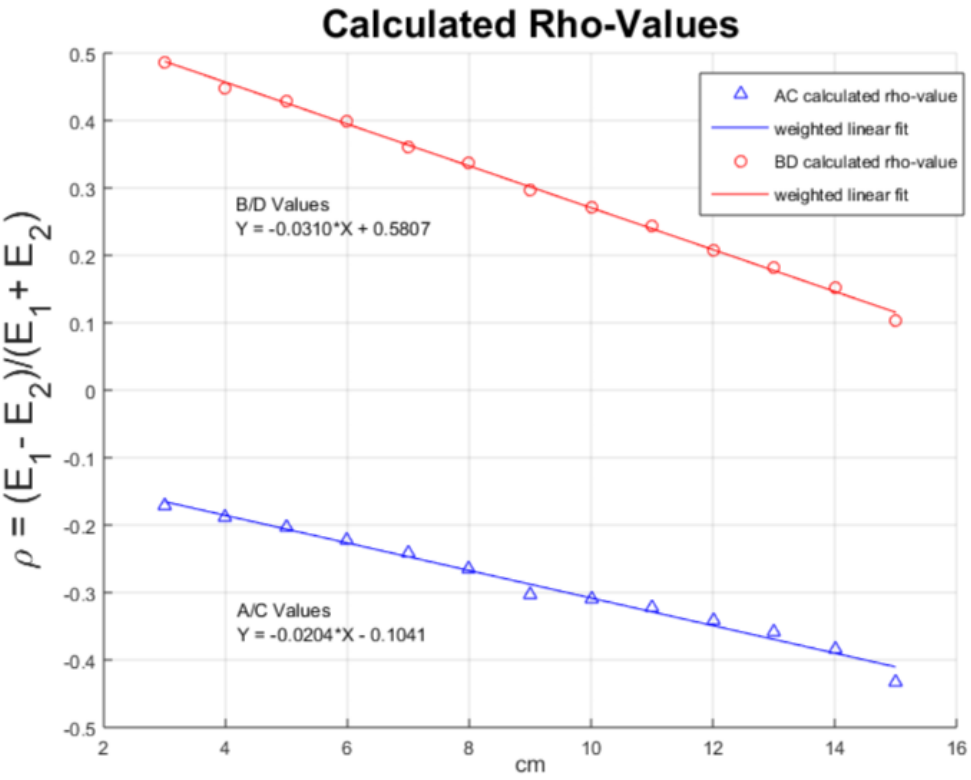

Figure 3: rho-value versus position for the two-scintillator setup. Note that though the rho-values differ between the scintillators, they still closely follow a linear fit.

We also discovered that the effectiveness of the optical grease decays over time. Assembly of the setup included adding Saint-Gobain's BC-630 Silicone Optical Grease to the interface between the crystal and the PMT. Figure 4 shows the energy readings of the same scintillator for several days following the application of the optical grease. As shown, the calculated rho-value for any particular position decreases over time, eventually settling to a value after a few days. This sort of systematic error will need to be taken into account for future setups; an assembled detector may need to settle for a few days before good data can be taken from it.

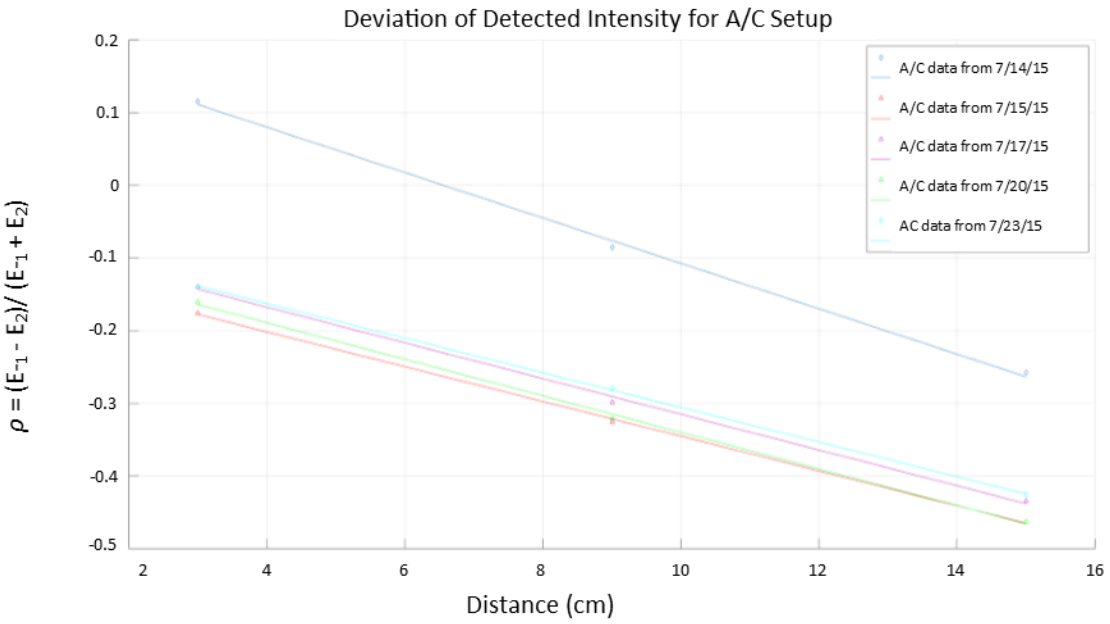

Figure 4: Calculated rho-values at three points (3, 9, and 15 cm) from the A/C setup on different days. The setup was assembled, and the optical grease applied, on 7/14/2015.

CHAPTER IV:

CONCLUSIONS

This work does well at being a simple proof-of-concept for position reconstruction given this design. The energy detected by our setup did reasonably well at reconstructing the position of the source. It's reasonable to say that in a large-scale setup, positions in 3D space can be determined with more complex programs and multiple applications of the methods presented here.

In future experiments, a way of improving accuracy may be found using time-of-flight (TOF) measurements. With these, the times when signals are detected will be used to determine the distance of the source from the scintillator, and further increase the accuracy of the position reconstruction. TOF methods were considered for this research, but the scintillator was too small for any useful difference in time to be measured. This variable will be more apparent in large-scale experiments. For useful TOF data, an oscilloscope with nanosecond timing will be necessary to precisely determine the difference in time it takes for each signal to arrive at its respective PMT.

Given these facts, the future of this technology seems promising. With better, more precise equipment, and more time, it seems possible to attain the accuracy necessary for full-scale medical imaging with this design.

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